Smart magnetic fluids:

Controllable distributed fluidic actuation

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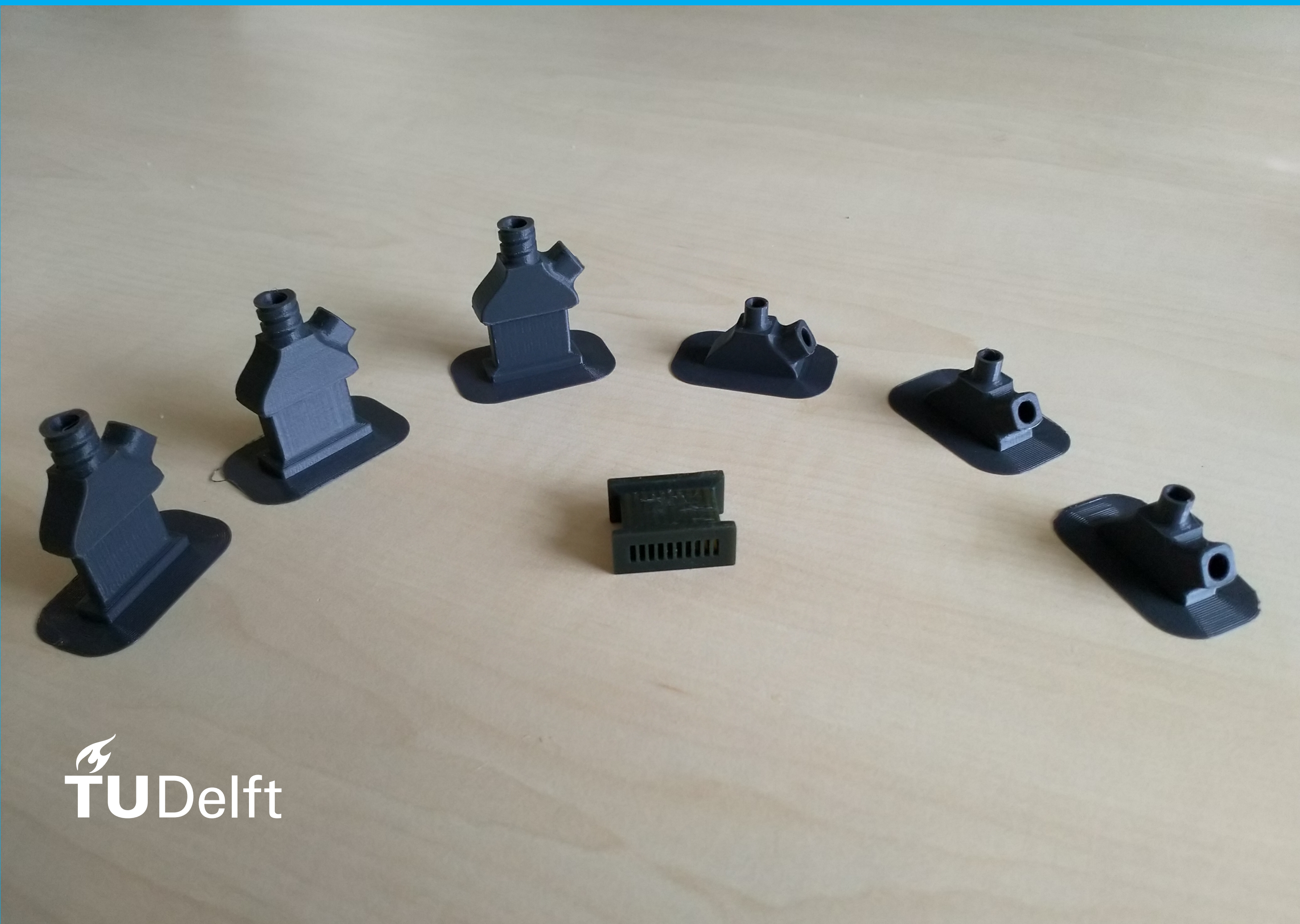

TU Delft

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Controllable distributed fluidic actuation

by

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TU Delft

Preface

On this page I want express my gratitude to all who helped me to finish my master thesis. In particular I want to thank Andres for his unwavering support with answers to my questions and discussions about the best course of action for the thesis. Secondly I would like to thank Hassan for all the feedback and the support for my project. I also wish to thank the lab support, specifically Patrick for the support in the design of the experiment and the practical tips and tricks for the experiment. Lastly I want to thank my parents, my brother and my friends for the good times beside my thesis, they always seemed to find a way to make me feel just a little bit better.

T.A. Strebus
Delft, September 2020

Abstract

This thesis is centred around the problem that heart failures become more common when other diseases could be treated better. Although surgery could fix the immediate effects of heart failures, the recovery after the surgery is hindered by the heart not being able to pump enough blood. There are solutions available but they are not implemented due doing more harm then good to the heart. This thesis will look into a way to aid the recovering heart by applying a compressional force to the outside of the heart. Previous attempts on fixing this problem could not follow the motion of the heart close enough due to the lack of actuation resolution, i.e. the local differentiation of the actuation applied to the heart. This caused damage to the heart muscle, which in the end causes death due to too little blood flow.

Due to lack of actuation resolution of the previous solutions, the focus of this thesis will be on the distribution of the actuation and not the actuation to the heart itself. A baseline for the requirements was needed to make an informed decision about the different solutions to distribute the actuation. This was done using a MatLab script. This gave the actuation characteristics of a compression of 20 percent with a pressure of 20 [kPa] when a compression jacket of 5 [mm] thickness around the heart is used. The relative high compression together with the pressure directed the solution for such an actuator towards fluids.

Common fluid actuators that would have a high local differentiation in actuation require lots of pumps or lots of valves, both of these options have as a downside that they take up to much space. That is where smart fluids have an advantage by decreasing the size of a pump or a valve by including the fluid in the active volume. Two of these smart fluids that seem promising are; Ferrofluids and magneto rheological fluids. Ferrofluids are activated by a magnetic field and will cause a change in internal pressure similar to how gravity affects internal pressure in a fluid. The only difference is that it could be about four to five orders of magnitude stronger over small distances of about a tenth of a millimetre. The magneto rheological fluid changes its viscosity in the direction perpendicular to the magnetic field with the change being related to the strength of the applied field.

The thesis will discuss the use of Ferrofluids as an active fluid in a pump, and magneto rheological fluids, MRF, as the active fluid in a valve. Ferrofluids are not feasible to be used as individual units as they barely meet the requirements of 20 [kPa] of pressure with the flow rate needed to actuate the jacket, without taking the fluidic losses of such a jacket into account. Whereas the MRF based valve could achieve a pressure difference of 20 [kPa] over the valve, without being of the limits of the operational value of the fluid. This is enough to have both a fully opened valve and a closed one using the pressure needed from the requirements. Therefor the thesis recommends the further research into implementing a full actuation system for a heart assist device using valves based on MRF.

Table of symbols

Symbol	description	value	unit
$\dot{Q}$	Heat flow		
$C_{p,b}$	Thermal capacity of blood	3.4	[kJ/K/kg]
ΔT	Temperature difference		
V_{mean}^{*}	Average flow rate of blood	70	[mL/s]
ρ_b	Density of blood	10^3	[kg/m^3]
$\mathbf{f_{int}}$	Internal volume force vector		
μ_0	Permeability in vacuum	$4 \cdot \pi \cdot 10^{-7}$	[H/m]
M_{sat}	Saturation magnetisation of the ferrofluid		
$\nabla\mathbf{H}$	Gradient of the magnetic field strength		
ΔP_{σ}	Pressure drop in MRF caused by H-field		
C	Factor dependent on viscous friction	2-3	[-]
σ	Yield stress of an MRF		
L	Length		
P	Mechanical power output of ferrofluid pump		
d	Maximum thickness of the ferrofluid layer		
Δp	Pressure difference		
A	Area		
t	Time of one stroke		
$M(H)$	Magnetisation of ferrofluid, dependent on H-field		
H_2	Maximum value of the H-field		
H_1	Minimum value of the H-field		
k	Geometry factor	48-96	[-]
μ	Kinematic friction		
D_h	Hydraulic diametre		
ρ	Density of the medium		
V	Average velocity of the medium		
H	Magnetic field strength		[kA/m]
t_c	Thickness of the coil		
w	Width		
j	Current density		[A/m^3]
g	Gap size, height of the channel		
ΔP_{η}	Pressure drop due to viscous friction		
η	Kinematic friction		
Q	Flow rate		
P_E	Electric heating power		
C_{cf}	Current needed for threshold over total current		
ρ_{Cu}	Resistivity of copper		

Table 1: List of symbols in order of appearance in the report

Contents

1 Introduction

Heart failure is a major health problem in both Europe and North America, with a patient population in 2007 of 10 million and 5 million respectively [1]. The heart will be more likely to fail with higher age, therefor due to an improvement in medicine, heart failures will become more frequent due to the increase in life expectancy in developed countries [1] [2]. Although medicine has increased the likeliness of success of the procedure to cure acute heart failure, the recovery after the surgery still have a relative large change of failing, i.e. a high mortality rate [1] [3]. This is due to the heart muscle being weakened after the surgery and thus having a lower ejection volume. This lower ejection volume of blood slows down the recovery or even worsens the situation by weakening the heart muscle even more. Which in the end leads to a heart that could not pump enough blood trough the body, which leads to death. Research has shown that patients of acute heart failure that did not have the lower ejection volume had lower mortality rates[3] [2].

Therefor a device is needed to keep the ejection volume, or the amount of blood per stroke at the normal level. This could be achieved by compressing the heart with a device that contracts the circumference of the heart. A device that performs this function and keeps the damage to the surrounding tissue does not exist, due limitations of the current actuators that are proposed. These actuators would damage the tissue too much due to the lack of fidelity. Therefor a new device is needed, this research will focus on finding and exploring new actuation options for a novel heart assist device for an increase in ejection volume of the left ventricle, while following the natural motion of the heart. Therefor a device is needed where individual segments of the actuator could be actuated to avoid pinch or pressure points on the heart that could damage the muscle, while being able to support and aid in the natural motion of the heart.

The heart assist will have to consist of distributed actuation due to the nature of the motion of the heart. This distributed actuation could further be applied to aid the weak spots in the heart's muscle. To find the right actuator for the heart assist a set of requirements had to be devised in order to remove the concepts that could not work from the ones that could. The concept for actuation and these requirements are addressed in the chapter 2. There are several ways to actuate an heart assist of which the most promising seem fluidic actuation or electromagnetic actuation due to their high energy and power densities. The proposed solutions that use these actuation types are patented but did not make it into products or did not work good enough. Due to the promising power densities, this thesis will look into a way of combining the two. Therefor the focus for this thesis will lie in smart magnetic fluids, i.e. fluids that change properties under the application of magnetic fields. The fundamentals for these fluids be looked at in the chapter 3, where actuators based on ferrofluids and magneto rheological fluids, or MR fluids, will be compared to the requirements. The ferrofluid solution is based on the idea of miniaturising a pump to supply each segment with its own pump, this is addressed in chapter 4. However, the ferrofluid based pump did not have a high enough power density and did not meet all the requirements. The MR fluids were analysed further in the chapter 5, where the MR fluids were utilised to redistribute the fluid flow from a single pump to several segments of the actuator. Each of this segments could have a pressure differential with respect to the pump pressure, which is controllable with an electro magnet. Therefor the problem statement is as follows:

"Could smart magnetic fluids provide a solution for creating localised actuation in a limited volume"

Bibliography

[1] E. B. Friedrich and M. Böhm, "Management of end stage heart failure," *Heart*, vol. 93, pp. 626–631, may 2007.

[2] D. L. Mann and M. R. Bristow, "Mechanisms and models in heart failure: The biomechanical model and beyond," *Circulation*, vol. 111, pp. 2837–2849, may 2005.

[3] I. Bytyçi and G. Bajraktari, "Mortality in heart failure patients," *Anadolu Kardiyoloji Dergisi*, vol. 15, pp. 63–68, jan 2015.

2

Heart assist device:
as seen for the perspective of an mechanical engineer

A translation between a medical and a mechanical view is necessary, in order to get the requirements for a heart assist device. Parameters that are useful in a medical field are not directly implementable in a mechanical engineering project due to their use case. A mechanical parameter should relate to the workings of the mechanism whereas the medical parameter should be directly measurable and comparable to existing or historic parameters. There are three parts needed in order to estimate the requirements; first the motion of the heart and most important intrinsic parameters of the heart, secondly the space available and the environmental constraints which were formed by means of an expert interview at the Amsterdam UMC and lastly the force and displacement generated by the heart muscle in actuation, which puts all the information from the previous sections into a script to build a computational model of the heart.

2.1. General motion of the heart

The movement of the heart can be described with two distinct motions. One being the shrinking of the radius of the heart and the other being a twisting motion. This is also seen in the direction of the muscle fibres as depicted in figure 2.1. All the fibres align in multiple twisted figure-8's where the two holes of the 8 overlap in the longitudinal axis of the heart, and the w-direction in figure 2.1. The x is distributed around the heart. Although both directions could be separated it is nice to think of them a one motion due to the continues nature of the fibres.

Modelling techniques could be applied to get a mathematical description of the fibres [A2]. To take into account that each heart is different, a medical ultrasound could be used to get the direction of the fibres and thus the motion of an individual heart [A1]. A motion is something dynamic and thus requires a time element. The heart rate is not the only parameter that determines the time it takes for the muscles to contract. The compressing motion is performed in about 40% of the time of a heartbeat with this fraction becoming slightly larger at higher heart rates. If the heart rate is 60 bpm the time in which the motion would occur is 0.4 seconds [A3]. Although the heart rate could rise way above this metric, it is quite a common value for the rest heart rate.

2.2. Expert interview

An interview was conducted at the Amsterdam UMC with two Anaesthesiologists after which multiples heart surgeries were followed to get a real life view of everything that was discussed. The infromation from this section is based on this visit and interview. The environment in the human body and in particular around the human heart was discussed with both with the anaesthesiologists. Due to the big differences per patient an general estimation was made to get a guide for the amount of space. The heart could be approximated with a cylinder with an outer radius of 5 cm and a height of 5 cm. This cylinder is capped at one end with half of a sphere again with a radius of 5 cm. The other end of the cylinder is capped with a closed wall. In reality a heart will differ in shape and size, but this base model will give a starting point with a quite average size and shape for the two ventricles of the heart. Furthermore the space available on the outside is around 3 cm around the heart.

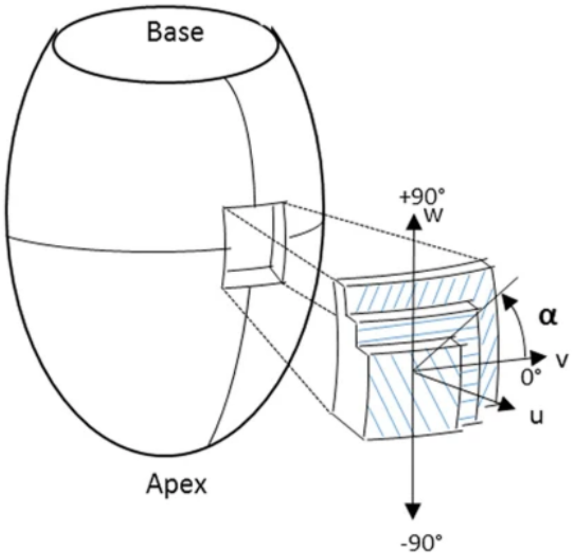

Figure 2.1: direction of heart fibres differs across the thickness of the heart muscle. Adapted from [A1]

The environment outside the heart has a very consistent temperature of around 37 degrees Celsius, but a lot of added heat could cause problems to functioning of the surrounding tissues. A quick way to find how much heat could be dissipated, is to assume that the flowing blood take away the heat. With a flow of around 70 mL per second (V^*_{mean}) and a thermal capacity of 3.4 kJ per Kelvin per kg ($C_{p,b}$). With a density of around 1000 kg per cubic meter (ρ_b).

$$\dot{Q} = C_{p,b} \cdot \Delta T \cdot V^*_{mean} \cdot \rho$$

This gives a cooling capacity ($\dot{Q}$) of around $(3.4 \cdot 10^3) \cdot 1 \cdot (70 \cdot 10^{-6}) \cdot (1 \cdot 10^3) = 2 \cdot 10^2$ Watts at a one degree increase in temperature (ΔT) of the blood, which is within the normal range. This assumes that all the heat is absorbed by the blood flowing through the heart, which is implausible, but if 5 percent of this value could be absorbed it is still about 10 Watts. Further research is needed to find the exact value, but this value of 10 Watts is an educated guess.
An last point of concern is that the tissue could be inflamed when a shearing motion is applied to the interface, therefor the interface between the heart and device has to be stationary with respect to the heart. A suitable solution for fixing the actuating surface to the heart should be found but is outside the scope of the research.

2.3. Computational model of the heart

A model of the heart is used to find the right force in the heart for different sizes of the heart. This model consists of a set of equations to uses the force produced by the muscles to find the corresponding blood pressure from which the flow rate is determined using the vascular resistance. This process is done iterative due to the blood pressure changing when the heart is contracting. The maximum force of the heart muscle varies quite a bit from person to person [A4]. Luckily the displacement is a lot more consistent across different persons and will make the final range a lot smaller. The displacement has an average maximum value of about 20 % shortening of the muscle [A3]. To avoid extensive hand calculations, a simple MatLab script is written in order to make the values that could be measured in a hospital comparable to values that are of interest for a mechanical engineer. With little adjustments this script could even be used to test future solutions for feasibility in a back of the envelope manner. For this rough estimation the heart is represented by a cylinder of muscle with a disk of blood in the middle of the cylinder, this method of modelling could be called a computational phantom heart. Both the wall thickness of the cylinder and the radius of the disk are parameters used for this equation. The strength of the muscle is taken from medical research [A4].
To simplify the workings of the script; it uses the geometry parameters, i.e. wall thickness and radius, to calculate the blood pressure at one moment. Secondly the blood pressure, as seen in figure 2.2, is

used to calculate the flow that would be achieved by using the estimated resistance from the vascular system. This flow is integrated of a small time step to get the new volume of the disk and therefor the radius. Then the new blood pressure is calculated and the loop starts again, which is repeated until the preset contraction of 20 percent is reached.

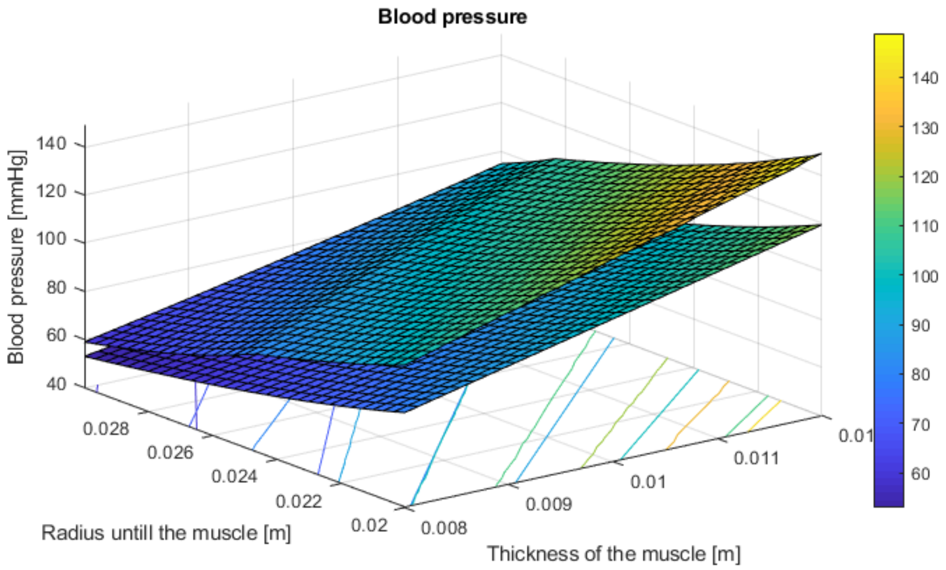

Figure 2.2: Blood pressure of the phantom heart at 7 cm length. The radius up to the muscle is at one axis and the other is the thickness of the muscle. The pressure is in mmHg to have a better comparison to the medical parameters.

In the end the start and end volumes are subtracted to get the volume of one stroke. Together with the time elapsed for one stoke the mean flow is calculated which could be seen for the 7 cm phantom in figure 2.3. Both the pressure on the beginning and at the end are noted. Furthermore the end time in the integration time steps is used to get the 'heart' rate seen in figure 2.4.

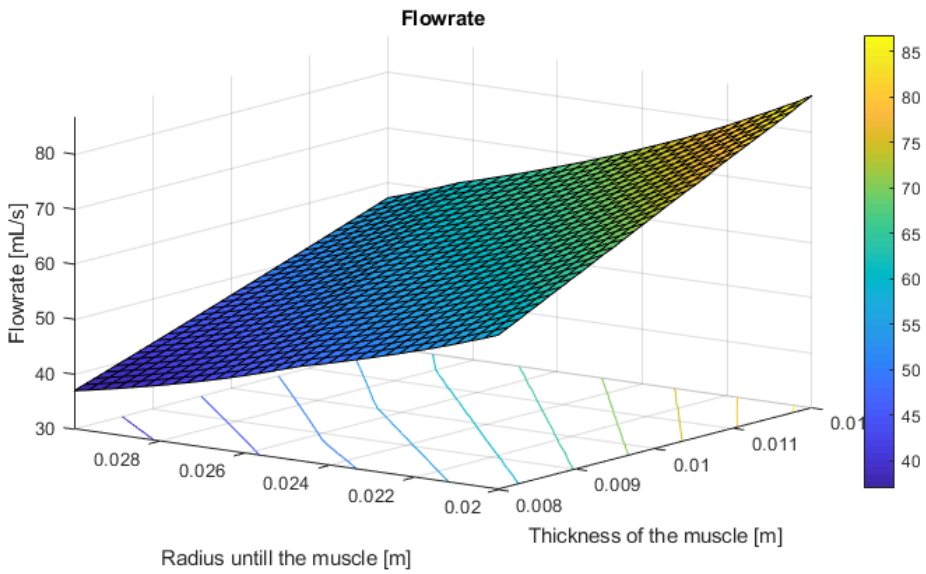

Figure 2.3: Average flow rate of the phantom heart at 7 cm length. The radius up to the muscle is at one axis and the other is the thickness of the muscle.

A few simplification are made to get a useful result in the minimal amount of time, both in run-time of the script and the modelling required to get closer to a real life situation, these simplifications will be reviewed in the discussion. The data produced has three geometry parameters namely, wall thickness

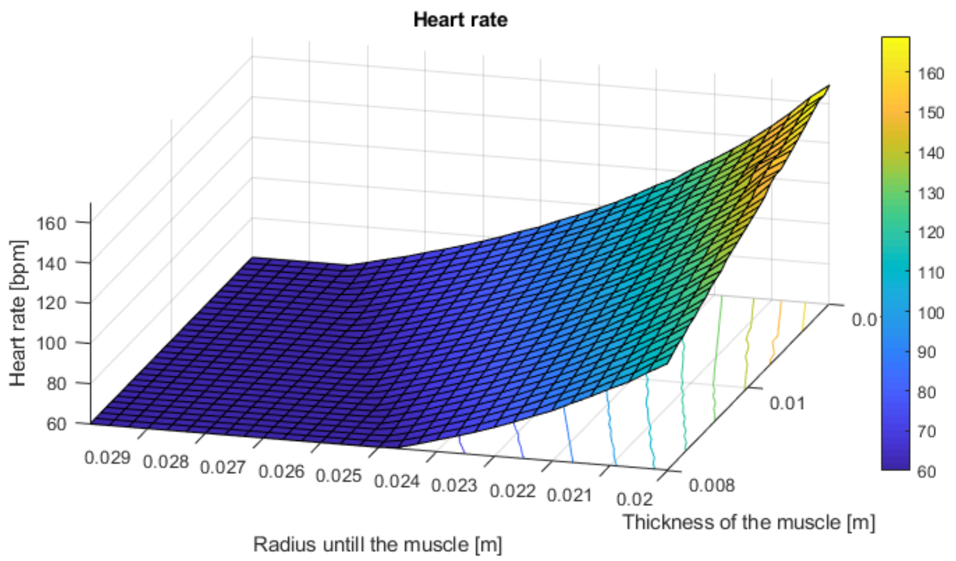

Figure 2.4: Heart rate of the phantom heart at 7 cm length. The radius up to the muscle is at one axis and the other is the thickness of the muscle.

of the muscle, the radius to the muscle and the total height of the cylinder. The figures are composed of one slice of that data taken from the data cube, to avoid clutter, this is done at a value of 7 centimetres for the height of the cylinder.
A constant force in the muscle fibres was used to calculate the blood flow that induced by this contraction. If the 20 percent contraction was not reached at 60 beats per minute, the values were marked and an additional force in the form of a compressing ring around the heart was added. The pressure needed in that ring is seen in figure 2.5. For now the heart rate was chosen as the targeting parameter but this could be changed in future revisions.

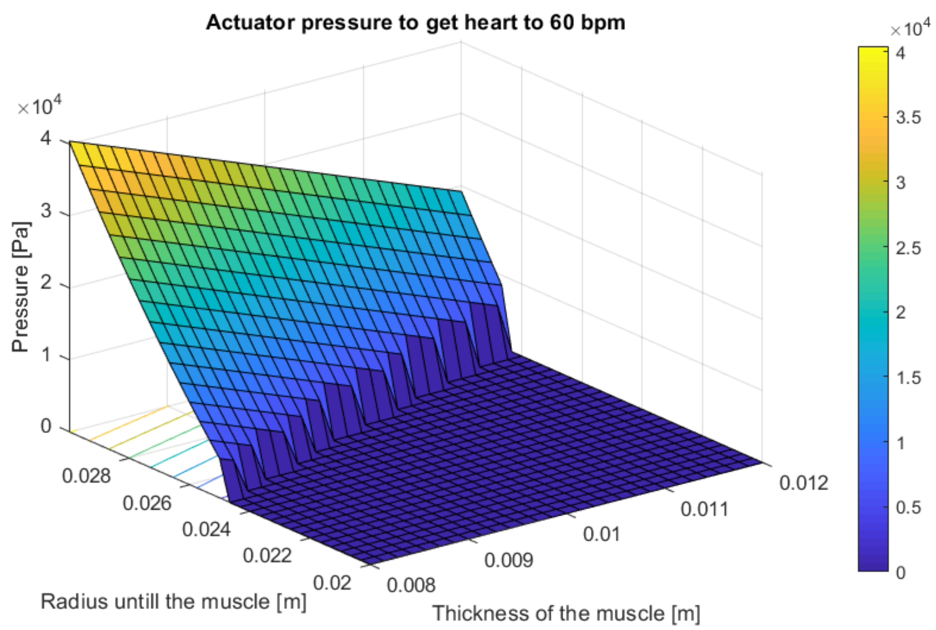

Figure 2.5: If the maximum contraction was not reached a extra force was calculated for the phantom heart at 7 cm length. This force is modelled as a compressive pressure at the outside of the muscle along the entire height of the phantom and with a thickness of 5 mm. The radius up to the muscle is at one axis and the other is the thickness of the muscle.

2.4. Summary

In order to aid the heart, the natural motion should be followed. This could be identified and modelled using 3D medical ultrasound and numerous model techniques. For the full motion of the heart, a compression in the muscle of about 20 percent is needed. Depending on the the shape and size of the heart and the force a heart can still deliver, the additional force needed can vary over multiple magnitudes. If an average sized heart is taken into consideration with about 70 percent of the regular force needed to actuate the heart in rest, the additional pressure needed to compress the heart, with a 5 mm thick actuating shell, is about 20 kPa. All the components of such a system should be fitted into the chest to avoid openings in the body. If the produced heat stays under 10 Watts, it is likely that no further cooling is required.

2.5. Discussion

The most significant simplifications are the shape of the modelled object, the exclusion of direction of the fibres, and the use of a simplified model of the vascular system that directly and linearly relates the pressure to a certain flow rate. With the need for a more elaborate model, these simplifications could be addressed with ease.

Paper heart References

[A1] C. Papadacci, V. Finel, J. Provost, O. Villemain, P. Bruneval, J. L. Gennisson, M. Tanter, M. Fink, and M. Pernot, “Imaging the dynamics of cardiac fiber orientation in vivo using 3D Ultrasound Backscatter Tensor Imaging,” *Scientific Reports*, vol. 7, pp. 1–9, dec 2017.

[A2] J. Wong and E. Kuhl, “Generating fibre orientation maps in human heart models using Poisson interpolation,” *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 17, no. 11, pp. 1217–1226, 2014.

[A3] J. Feher, *Quantitative Human Physiology*. Elsevier, 2012.

[A4] E. H. Sonnenblick, W. W. Parmley, R. A. Buccino, and J. F. Spann, “Maximum force development in cardiac muscle [20],” 1968.

3

The feasibility of smart magnetic fluids in a hydraulic actuator

Abstract

This paper will discuss potential actuating principles for a distributed surface actuator. Fluidic actuators like McKibben muscles are already used in medical application where a positive pressure has to cause a contraction like a muscle. These actuators have the big downside that the complexity, and thus the total size, goes up when multiple of these sectioned actuators would be used to create a distributed surface actuator. To avoid this smart magnetic fluids could simplify the total system by providing a solution of an individual pump per segment with ferrofluids or a variable valve using magnetorheological fluids.

3.1. Introduction

An expert interview, with anaesthesiologists from the Amsterdam UMC, revealed the need for an distributed surface actuator to aid the heart muscle. Another application in the same field could be to aid the digestive track. This paper will focus on the exploration of different actuation techniques that meet the requirements that are set in the paper in chapter 2. At this moment most actuation is done by actuators in bulk, i.e. each motion is performed with a fixed set of actuators. Actuation in bulk has different drawbacks including size and the lack of local differentiation of the actuation. There are multiple ways of overcoming this drawback, for example, use smaller novel actuators or use an extra layer to distribute the actuation. Both have their advantages and disadvantages. Both have the advantage above the bulk actuation of the local actuation due to the increase in fidelity.

This research will focus on the technical aspects of a potential actuator to be used in a distributed surface actuator. Primarily the type of actuation that would be suited for such an application. As mentioned earlier the compliant distributed surface actuation is missing in current concepts in aiding the heart muscle in the human body. Where the compliant part refers to the need for the device to be formed to the environment while maintaining its functionality. The differentiation of local deformation according to the current situation is covered by the distribution part. The research will therefor put its focus on one of the strongest complex organs in the human body; the heart. This will be taken as the benchmark to set the bar high and secondly the heart combines the compliant and distributed part in great extend. With the requirements detailed in the paper in chapter 2, together with the goal of the research will guide the search through literature towards Variable mechanical Impedance Actuators [B1] or in short VIA. The need for compliance pushes the research towards fluids, a very comprehensible review by Hines et al. [B2] gives a welcome overview about a big portion of the existing actuation techniques. Current solutions will mostly be focused on fluid actuation with compliant McKibben muscles or similar fluid actuators [B3] [B4]. The VIA will give the foundation to start the search in the literature and will be discussed in section 3.2. A big drawback these fluid actuators share is the need for external devices, such as valves and pumps, which becomes a large problem if a surface of these actuators is needed and with every increase in linear resolution, the need for more valves or pumps scales quadratic. With the overview in mind there are two solutions that seem promising to solve the problem by reducing the parts needed;

- Ferrofluids as a pump where current is related to the pressure created by the pump and thus removing the need for valves. This solution will be discussed in section 3.3.
- Magneto rheological fluids as a pressure reducing valve and thus creating a small footprint valve that directly links current to a pressure drop. A more detailed explanation could be found in section 3.4.

3.2. Variable impedance actuators

As the field of robotics and mechanisms evolves there are new possibilities for the construction of mechanisms. One of these possibilities lies in the mimicking of biological mechanisms where the actuator is intertwined with the support structure and sensing. Although sensing is outside the scope of this research, it will be important to be looked into in future work. The need for the combination of structure and actuation is, if it is put it very black and white, because humans or biological creatures are soft and can change the way they respond as a system to the environment, robots or mechanisms are hard and or not capable of change. Therefor if the same flexibility to a changing environment is needed, without over-designing a system, the functions of actuators and support structure should be combined. To capture this in a single term, actuators with 'variable mechanical impedance' are one of the solutions for the lack of flexibility in current mechanisms. In this way of seeing actuators, the mechanical impedance refers to the stiffness and damping of a system.

State of the Art

In the state of the art we see many systems that are capable of changing the impedance of the system, but they are not compliant. The changes of the system are mostly focused on the actuator and specifically on the compliance of the actuation, or in other words the systems has a build in conservative energy storage. The systems use solutions that do not scale down easily therefore these systems are generally bulky and thus not the type of systems that are needed in the human body. Some systems are focused of the change in damping force, or systems that could dissipate energy out of the system [B5][B1]. This seems less useful because energy will be lost. This type of systems still be useful because the effects scale down quite well and thus will produce smaller actuator systems. Although in this category there are systems whose characteristics are in the right direction for actuation inside the human body, for example fluid dampers and plastic deformation dampers, there still is the need for an external individual actuation systems which makes both not viable. To be more specific, fluid dampers need a valve to control the damping and the plastic deformation dampers need an secondary systems to 'reset' the deformation in order to use it consistently. Both of those external actuators will make the total system bigger and with more non-compliant parts.

Towards distribution

To achieve a more distributed actuator that what is done in the state of the art, the scope was set on magnetic fluids, to combine the compliance of the fluids and the 'in-situ' changes of the system with the relative ease of a change in magnetic field. From the overview of different actuation types [B2] and the requirements, from my paper in chapter 4, two potential candidate arise namely Ferrofluids, in short FF, and Magneto-Rheologial Fluids, or MRF. Their electric counterparts will not be considered due to the high electric potential not being suitable inside the human body. Ferrofluids will be discussed in more detail in a separate paper in chapter 4. The MRF will be reviewed in the section 3.4 due to the interesting property regarding their change in viscosity with a magnetic field applied. Both fluids have the ability to be adjusted locally and thus making a more distributed actuator possible. The two fluids represent would have a different way of achieving the distributing of the actuation. For the ferrofluid this is more linked to the change in pressure and thus the stiffness of the actuator. It could be envisioned as a pump that could drive each individual segment of a distributed surface actuator. If the MR fluid is guided through a channel to which a magnetic field could be applied it is a drop in replacement for a valve. A pressure drop could be produced in a channel by the local change in viscosity induced by the magnetic field.

3.3. Ferrofluids

The ferrofluids could change the internal fluid pressure, simular to how gravity increases pressure in a fluid [B6]. The difference between the two, is that with gravity, the pressure distribution will follow the gravitational field and with the effect in ferrofluid will act along the magnetic field lines. The control over the internal pressure would be very beneficial for an actuator that has to change its impedance, due to the pressure being linked to the stiffness of the fluid in a actuator.

State of the art

At this moment there are multiple uses for ferrofluids, one of these is the use as a fluid bearing [B7]. Another use is an actuator, because the ferrofluids increase the local fluid pressure it could be used to move objects [B8] [B9], which could also be the ferrofuid itself as shown in a atmospheric disturbance correcting mirror [B10]. These examples also show one of the downsides of ferrofluids as an actuator; the stroke that could be achieved is relatively low, in the order of millimetres.

Simulations

In a simulation there is seen that the pressure close to the magnetic source is high and quite usable, if the distance is further from the magnetic source, the effect is not very usable because the forces would be lower than could be achieved with other means, as seen in paper in chapter 4. The same effect could be seen in ferrofluid seals [B2]. In other words the energy density of the actuator gets lower if the stroke needs to be bigger, this could be solved by keeping the stroke low and the actuator bigger by area perpendicular to the stroke. This is limited by the volume needed in order to generate and guide the magnetic field, i.e. when the area is larger the volume needed to provide a large enough magnetic field increases too.

Internal Ferrofluid pressure The force of the ferrofluid is an internal one, the particles are pulled together by the individual magnetic domains that are aligned to the magnetic field lines. The higher the absolute value of the magnetic field, the higher the internal force which results into a pressure. The internal volume force could be calculated by [B6]:

$$\mathbf{f_{int}} = \mu_0 \cdot M_{sat} \cdot \nabla\mathbf{H} \tag{3.1}$$

Where $\mathbf{f_{int}}$ is the vector of the internal volume force, μ_0 is the permeability of vacuum, M_{sat} is the saturation value of the magnetisation of the used ferrofluid, and lastly $\nabla\mathbf{H}$ is the vector of the slope of the magnetic field. The magnetisation of the ferrofluid is dependent of the magnitude of the magnetic field, for the high fields that are required for a usable pressure for an actuator, the bulk of the fluid is in saturation.

Maximum realistic pressure In Ferrofluid seals the pressure in the fluid could reach 1 to 2 bar depending on the fluid and the magnetic circuit [B7] [B11] [B6]. For an actuator this is about one order of magnitude lower, as seen in the paper in chapter 4. Although it is possible to get closer to the pressure of the FF seal, this is only in a more theoretical case with little to no losses. If a more sensible estimation would be that the pressure reached is about 0.1 [bar]. This is close to the 0.2 [bar] from the requirements taken from my own paper in chapter 4.

Power density An actuator is characterised by many parameters, but if it is designed for an enclosed space, the energy density is one of the most important. The FF actuators seem promising with a relatively high pressure, this is, however, paired with a relatively low stroke. All of the aforementioned parameters are compared with a biological muscle. The parameter that helps the FF actuator is that it is fast. A rough estimation puts the power density in the neighbourhood of 1 Watt per litre as seen in the paper in chapter 4. To give this perspective, a heart in rest has an output power of about 1 Watt [B12].

Conclusion and recommendations

Ferrofluids seem promising to be used as actuating fluids. A usable pressure could be achieved, but when losses are included this might not, or might just be enough to support a heart in rest depending on the level of support needed. Therefor no further research is done regarding the losses in a ferrofluidic actuator. A ferrofluidic actuator would make sense in a situation where the magnetic circuit is already there, for example active balancing of a rotor in a electric motor, or if the ferrofluid is used as a bearing and a small adjustments to the height are needed, e.g. in a sliding platform for a microscope.

3.4. Magnetorheological Fluids

An magnetorheological fluid, or MR fluid, is carrier fluid, mostly oil based, with ferromagnetic particles suspended in the fluid. These particles are between 0.1 to 10 micron. Although fluids exist with bigger and smaller particles, this range is the most used because the bigger particles will settle more easy and the smaller particles will not have the wanted characteristics and will behave more like a ferrofluid [B13]. This section will focus on the working principle of the MR fluid and on the more practical application that have been already used. The word actuator is used in a more general sense of the word where it is more comparable to an transducer that will convert one kind of energy to another energy and vice versa. Therefor dampers are considered as actuators because they will convert mechanical energy into heat energy.

MR fluid is activated by an applied magnetic field where the iron particles align to this field. In this process, chains of iron particles will be formed that will increase viscosity. When the magnetic field is strong enough, the fluid can go to a gel like state and thus behave like a solid. The iron particles could be coated in a lipophilic coating to keep them from settling under gravity due to their higher density [B14]. Another solution is to use a filler to have a pseudo-structure in the fluid, a potential candidate could be fumed silica as shown by [B15]. This process of the change in viscosity could be seen in figure 3.1, where the bigger grey particles represent the iron and the smaller blue ones the fumed silica filler.

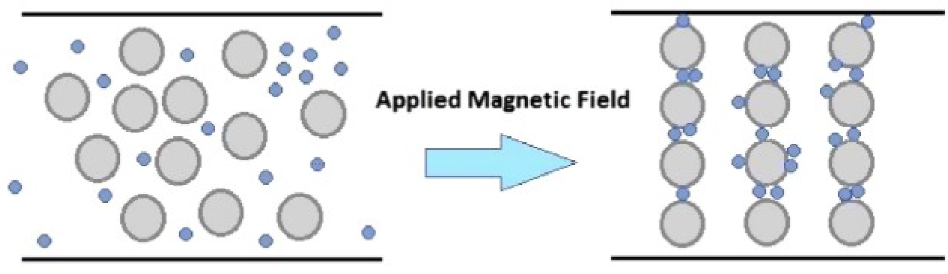

Figure 3.1: Chain forming of iron particles in a MR fluid under applied magnetic field. The field is applied in the vertical direction and due to the chain forming the flow is blocked in the horizontal direction. The bigger grey dots represent the iron particles and the smaller blue dots the filler. Adapted from [B14].

State of the art

MR behaviour categorisation

In this subsection the working principle of the MR fluid is the central theme. There are multiple models to describe the behaviour of the MR fluid, both macroscopic and microscopic, because of the size of the application in mind only the macroscopic models are taken into account. The microscopic models would be too computational intensive for a centimetre scale actuator, without a big increase in the accuracy of the model. The macroscopic models for the fluid are build with the microscopic behaviour in mind. Therefor all the assumptions in these models shine though as small errors in the macroscopic models. For an overview of the different domains a Pipkin diagram of ER fluid is used as seen in figure 3.2. The rheological behaviour of the MR fluid is similar as ER fluid [B2][B16]. The four regions are bound by a set of arbitrary chosen limits due to the continuous nature of the different models. For example the viscoplastic model will be equal to the Newtonian model if the frequency or the strain amplitude is large enough. Therefor the limit at boundary ”c” is set by the authors of [B16] as if the added viscosity due to the electric field, is within 10 % of the viscosity that it would have without the electric field, the Newtonian model could be used. For the limit between the linear and non linear viscoelastic model, boundary ”a”, it is set that the shear stress of the first harmonic is contributing the largest to the total shear stress or specifically for this figure 90 % of the total. Boundary ”b” is defined as the limit where the the elastic forces are just 10 % of the total and thus the plastic or viscous part is dominant. The goal for these flow regimes is to get a approximate idea for the different models and when to use them. In the following paragraphs the different models will be described in more detail and their relation to the MR fluid. The Newtonian model will not be discussed due to the limited applicability for the intended purpose for the MR fluid in this research, i.e. the effect of the added viscosity is not predominant in this region and thus not useful.

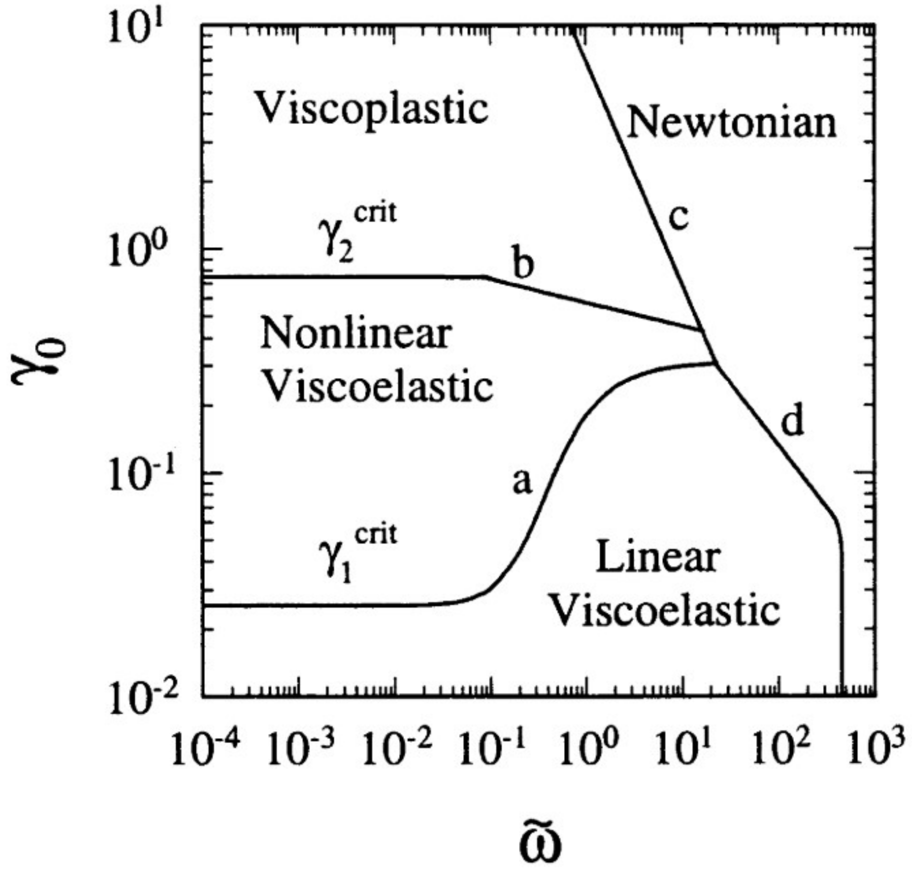

Figure 3.2: A Pipkin diagram of ER fluid, the different rheological behaviours are seen in this diagram. The y axis is the strain amplitude and the x axis is the normalised and dimensionless frequency. Due to the simularities between MR fluid and ER fluid, the same rheological behaviours are seen in MRF. Adapted from [B16]

Viscoplastic behaviour This behaviour could be described as a Bingham fluid with variable yield stress as seen in figure 3.3, this model is easy to implement and will give a relatively accurate prediction of the pressure drop in a fluid. This model is used in the next section, where it is implemented in MatLab, to get an estimate of the pressure drops and power usage. The Bingham model is a variation on the Newtonian fluid model, where a constant shear stress is added. For the MRF this shear stress is variable and dependent on the magnetic field strength. The Herschel-Buckley model is also seen figure 3.3, this model could, according to [B2] and [B16], be a better fit to describe the behaviour of a MRF. For the purpose of getting an estimate of the pressure drop a Bingham model is better suited due to the fact that some parameters of the MR fluid should be experimentally acquired to get a valid Herschel-Buckley model. For the Bingham plastic model these parameters are known.

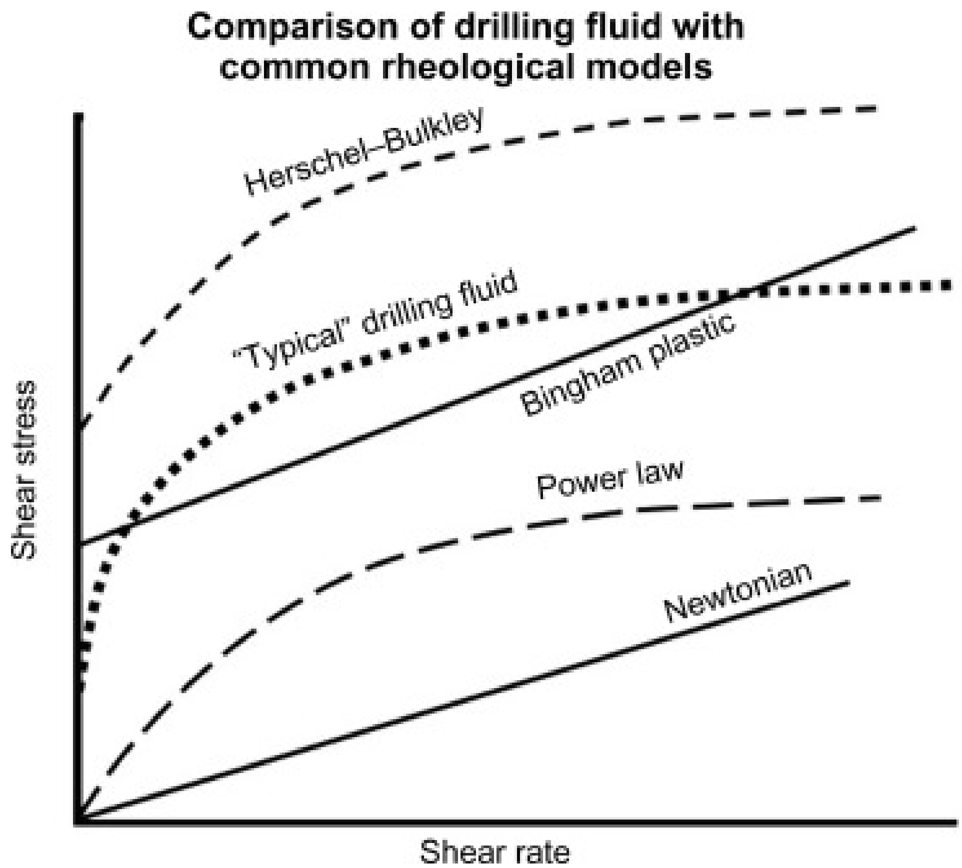

Figure 3.3: Different fluid models taken from [B17]

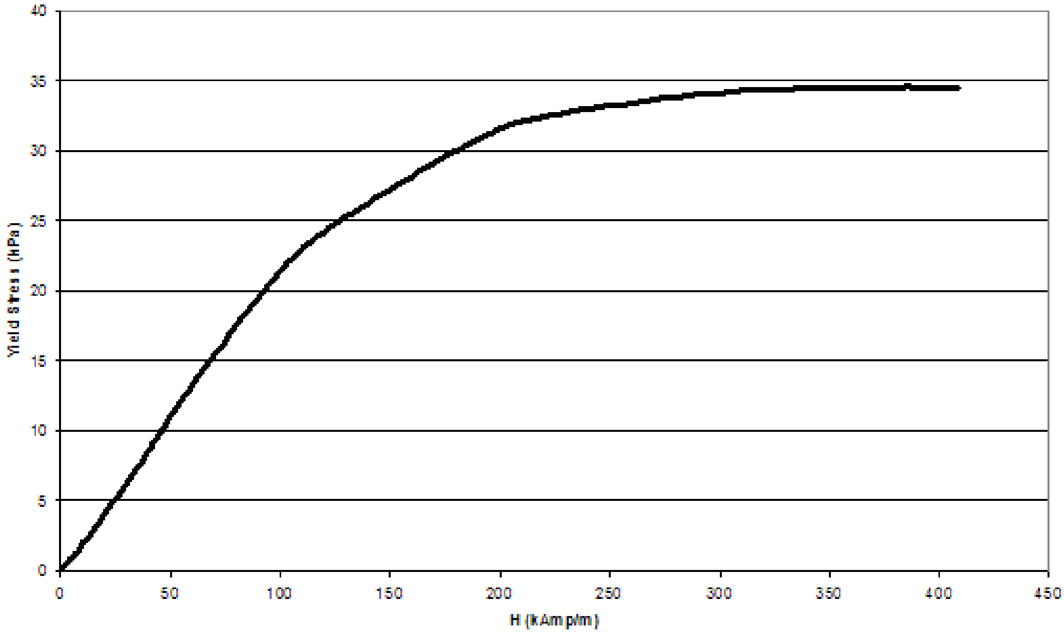

Figure 3.4: Yield strain versus magnetic field strength for the fluid MRF-122EG from [B19]

Other rheological behaviours As seen in the figure 3.2, under small amplitudes and small frequencies the MR fluid will behave with a linear viscoelastic response, i.e. the fluid will store part of the work applied to the fluid and the rest will be dissipated as if it was a pure fluid. When the frequency increases the stored part will be negligible to the dissipated part and thus a Newtonian model could be used. If the strain amplitude crosses the first critical boundary, noted by 'a' in the figure. Non-linear effects will dominate the stress strain relation, to model these behaviours would take a complete frequency, stress and strain sweep to characterise this behaviour. This will result in a complex model, both in characterisation and parameters, therefor the strain should be big enough to avoid this region and thus this complex behaviour.

Pressure drop estimation

The pressure drop over an actuator could be estimated using a simple formula from [B18]:

$$\Delta P_\sigma = \frac{C \cdot \sigma \cdot L}{g} \tag{3.2}$$

Where the pressure drop due to the application of the magnetic field, ΔP_σ, is related to a factor C, which is dependent on the difference between the viscous pressure and the pressure drop due to magnetic field. This difference is therefor dependent of the flow speed, but does not vary much, between 2 and 3. L is the length of the valve and g the height or gap of the flow channel, as long as the magnetic field is constant across the width of the valve it does not matter how wide it is. Further more σ is a value that is proportional to the magnetic field strength and dependent on the fluid. For relatively low fields, until 50 or even 100 [kA/m], a linear relation could be assumed for some commercial available MR fluids. This linear part could be seen in the figure 3.4, and could be expressed as $\sigma = \frac{21.5}{100} \cdot H$ for the MRF-122EG fluid from LORD [B19].

With the reference geometry of figure 3.5, formula 3.2 is then used to estimate the pressure drop. With a length L of 1 [cm] and a gap g of 0.2 [cm] and a field strength H of 10 [kA/m], which would result in a yield stress of 2.15 [kPa], the pressure drop will result in:

$$\Delta P_\sigma = \frac{2 \cdot 2.15 \cdot 0.01}{0.002} = 21.5 \text{ [kPa]}$$

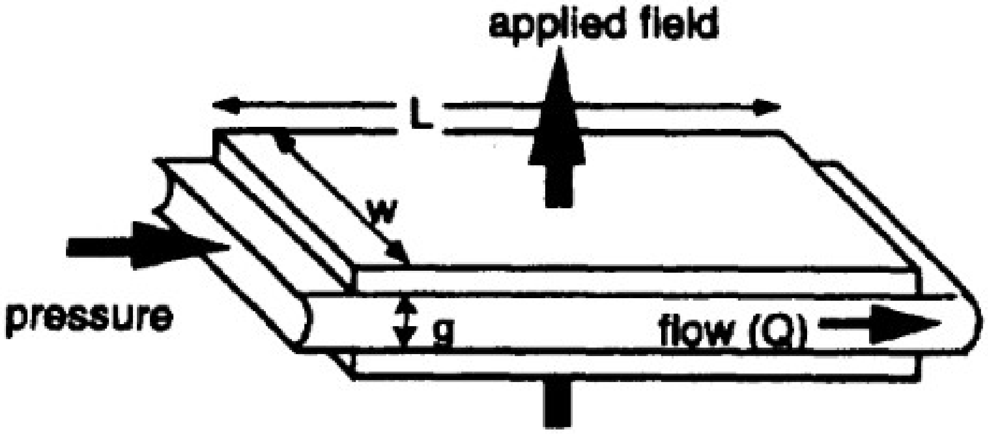

Figure 3.5: Reference geometry adapted from [B18]

Results and discussion

Even with a relatively low field and an fluid with a lower iron content by weight, and thus producing a lower reaction to the magnetic field an proportional valve should be possible with maximum blocking pressure of about 20 [kPa]. A electro magnet could be used to produce this field in the gap and thus make the valve proportional to the applied current. The viscous losses are not yet taken into account and should be calculated or simulated. The bingham model on which this calculation is based does not reflect the behaviour of the fluid perfectly, although for a rough estimation this is not a problem. A potential candidate for a better describing model of the behaviour is the Herschel-Buckley model, because it follows the behaviour of the MRF more closely. This is needed when the valve is controlled by a control system in a total system.

3.5. Conclusions

Both smart magnetic fluids, the MR fluid and the Ferrofluid, show potential to be used as an active fluid to control or actuate mechatronic systems.
For the application in mind the Ferrofluid is not suitable, because it barely meets the requirements without the losses taken into account. This is not to say that it could not be useful in other applications. The paper in chapter 4 will go into more detail about the operation range for a cm scale pump. To summarise the pump is estimated achieve a pressure of about 0.1 [bar] at a flow rate of 10 [mL/s] for a 4 [cm] by 2 [cm] by 1 [cm] at the theoretical operation limit of 100 [Hz].
A MRF based valve show even more potential by exceeding the requirements without operating at the theoretical limits. The valve would be able to produce a blocking pressure of about 20 [kPa] when a field of 10 [kA/m] is applied in a 1 [cm] long and 2 [mm] high flow channel. The width of the valve does not influence the performance of the valve as long as the field could be applied in a constant manner across the flow channel. Although there is the need for a single external pump, all the actuation could be split and controlled using the multiple MRF based valves. A more detailed estimation of the valve unit will be discussed in the paper in chapter 5

Paper LIT References

[B1] B. Vanderborght, A. Albu-Schaeffer, A. Bicchi, E. Burdet, D. G. Caldwell, R. Carloni, M. Catalano, O. Eiberger, W. Friedl, G. Ganesh, M. Garabini, M. Grebenstein, G. Grioli, S. Haddadin, H. Hoppner, A. Jafari, M. Laffranchi, D. Lefeber, F. Petit, S. Stramigioli, N. Tsagarakis, M. Van Damme, R. Van Ham, L. C. Visser, and S. Wolf, “Variable impedance actuators: A review,” *Robotics and Autonomous Systems*, vol. 61, pp. 1601–1614, dec 2013.

[B2] L. Hines, K. Petersen, G. Z. Lum, and M. Sitti, “Soft Actuators for Small-Scale Robotics,” *Advanced Materials*, vol. 29, p. 1603483, apr 2017.

[B3] A. De Greef, P. Lambert, and A. Delchambre, “Towards flexible medical instruments: Review of flexible fluidic actuators,” *Precision Engineering*, vol. 33, no. 4, pp. 311–321, 2009.

[B4] M. De Volder, A. Moers, and D. Reynaerts, “Fabrication and control of miniature McKibben actuators,” *Sensors and Actuators A: Physical*, vol. 166, pp. 111–116, mar 2011.

[B5] M. Catalano, G. Grioli, M. Garabini, Felipe Weilemann Belo, A. Di Basco, N. Tsagarakis, and A. Bicchi, “A Variable Damping module for Variable Impedance Actuation,” in *Proceedings - IEEE International Conference on Robotics and Automation*, pp. 2666–2672, Institute of Electrical and Electronics Engineers Inc., 2012.

[B6] R. Rosensweig, *Ferrohydrodynamics*. Mineola: Dover Publications, 2018.

[B7] A. S. Boots, L. E. Krijgsman, B. J. de Ruiter, S. G. Lampaert, and J. W. Spronck, “Increasing the load capacity of planar ferrofluid bearings by the addition of ferromagnetic material,” *Tribology International*, vol. 129, pp. 46–54, jan 2019.

[B8] E. Uhlmann and N. Bayat, “High Precision Positioning with Ferrofluids as an Active Medium,” *CIRP Annals*, vol. 55, pp. 415–418, jan 2006.

[B9] R. Olaru, A. Arcire, C. Petrescu, and M.-M. Mihai, “Study of the magnetic force delivered by an actuator with nonlinear ferrofluid and permanent magnets,” *IEEJ Transactions on Electrical and Electronic Engineering*, vol. 12, pp. 24–30, jan 2017.

[B10] D. Brousseau, E. F. Borra, and S. Thibault, “Wavefront correction with a 37-actuator ferrofluid deformable mirror,” *Optics Express*, vol. 15, p. 18190, dec 2007.

[B11] M. De Volder and D. Reynaerts, “Development of a hybrid ferrofluid seal technology for miniature pneumatic and hydraulic actuators,” *Sensors and Actuators A: Physical*, vol. 152, no. 2, pp. 234–240, 2009.

[B12] J. Feher, *Quantitative Human Physiology*. Elsevier, 2012.

[B13] M. T. López-López, P. Kuzhir, S. Lacis, G. Bossis, F. González-Caballero, and G. Durán, “Magnetorheology for suspensions of solid particles dispersed in ferrofluids,” *J. Phys.: Condens. Matter*, vol. 18, pp. 2803–2813, 2006.

[B14] M. Ashtiani, S. H. Hashemabadi, and A. Ghaffari, “A review on the magnetorheological fluid preparation and stabilization,” jan 2015.

[B15] S. T. Lim, M. S. Cho, I. B. Jang, and H. J. Choi, “Magnetorheological characterization of carbonyl iron based suspension stabilized by fumed silica,” in *Journal of Magnetism and Magnetic Materials*, vol. 282, pp. 170–173, North-Holland, nov 2004.

[B16] M. Parthasarathy and D. J. Klingenberg, “Large amplitude oscillatory shear of ER suspensions,” *Journal of Non-Newtonian Fluid Mechanics*, vol. 81, pp. 83–104, feb 1999.

[B17] M. S. Ramsey, “Rheology, Viscosity, and Fluid Types,” in *Practical Wellbore Hydraulics and Hole Cleaning*, pp. 217–237, Elsevier, jan 2019.

[B18] M. R. Jolly, J. W. Bender, and J. D. Carlson, “Properties and Applications of Commercial Magnetorheological Fluids,” *Journal of Intelligent Material Systems and Structures*, vol. 10, pp. 5–13, jan 1999.

[B19] LORD, “MRF-122EG Magneto-Rheological Fluid.” [online] Available at: http://www.lordmrstore.com/lord-mr-products/mrf-122eg-magneto-rheological-fluid.

4

The design principles and feasibility study of a Ferrofluid pump

Abstract

In this paper the feasibility of a novel pump based on ferrofluids is presented. Ferrofluids are used in many application ranging from seals, to valves, to positioning stages. With this study the possibility of a centimetre scale ferrofluid based pump is explored. With calculations and simulations the limits of such a pump are being found within the current specifications of existing ferrofluids. The pump should be able to work at a frequency of 100 [Hz] to displace 10 [mL/s] at 0.1 [bar], this includes an estimation for the viscous friction and an estimation for the magnetic field losses..

4.1. Introduction

Hydraulic actuators are great for actuating, the force density is high combined with an almost unlimited stroke. They do have one big short coming and that is the downscaling. Although the actuators scale fine down to centimetre level, the external devices needed, like valves and pumps do not, definitely if high fidelity is required, in this case more levels than just opened or closed. The high fidelity is comparable to audio, due to the discretisation of music the quality degraded or from a different perspective an error with regard of the original signal is formed. The higher the fidelity, i.e. the smaller the discretisation step, the better the analogue signal is represented. The same holds true for surface actuation, where the actuation of the surface should follow an predefined path. The different parts of the surface are discretised and thus the parallel between the audio and this actuation could be drawn. When fidelity is high the discretisation of the actuation follows the required path with less error. This would mean many small pumps or many valves in order to get to this high fidelity. Not only a high discretisation count is part of the high fidelity also the different levels between on and off. For valves this is on or off, while pumps could be controlled to a certain level, but the control is not direct due to the transfer from electrical to mechanical to another form of mechanical (fluid) energy. To increase the fidelity, multiple smaller pumping units should be used with a direct control of pressure by in this case electricity. This could be achieved with Ferrofluids, due to their ability to change the internal pressure in the presence of a magnetic field. This research will focus on the design principles and feasibility of such an actuator or pump unit.

4.2. Background

Ferrofluid are used in different fields due to their ability to react to a magnetic field while staying liquid. A few applications are deformable mirrors [C1], Ferrofluid seals [C2] [C3], micropumps [C4] and microvalves [C5], positioning stages [C6], magnetic field sensors [C7], a tactile display [C8] and many more [C9]. The existing design for different ferrofluid actuators can be divided into 2 main categories with multiple sub categories.

- Direct drive ferrofluid actuators, where the ferrofluid is used to actuate, i.e. a sort of pump. [C4] [C5] [C10] [C11] [C12] [C13] [C14] [C15] [C16] [C17] [C18] [C19] [C20]
- Indirect drive ferrofluid actuators, where the ferrofluid is used to push a non-ferrous material out of the way. [C6] [C21] [C22] [C23] [C24] [C25] [C26] [C27] [C28]

An indirect drive ferrofluid actuator could be described as a direct drive actuator with a mechanical “link”. Such a link could be a piston that is pushed out of the way by the ferrofluid, or another fluid or a membrane. Due to this reasoning the direct drive ferrofluid actuators are a good starting point, to develop a new actuator. If the overview is expanded to also include non-actuating ferrofluid applications such as bearings and seals, there is seen what kind of design has the most promising signs for a high power density, large stroke, small design. There is seen that with a small stroke the power density is quite high, a stroke of in the order of 0.1 mm with a pressure of in the order of 0.1 bar, this could be achieved in about 10 milliseconds [C29] [C30]. This will give a power of for 2 by 0.5 [cm^2] of area:

$$P = \frac{d \cdot \Delta p \cdot A}{t} = \frac{(0.1 \cdot 10^{-3}) \cdot (1 \cdot 10^{4}) \cdot (1 \cdot 10^{-4}))}{(10 \cdot 10^{-3})} = 0.01 \text{ Watt}$$

The space needed for the Ferrofluid is about 1 cubic centimetre, the coils needed to generate the magnetic field will take up about 2 cubic centimetre plus another cubic centimetre for the magnetic circuit and another 4 cubic centimetre to reduce magnetic losses, the whole system would take up about 8 cubic centimetre within a 1 [cm] by 4 [cm] by 2 [cm] enclosed volume. This would result in a theoretical actuator with a power density of

$$P_d = \frac{0.01}{8 \cdot 10^{-6}} = 1250 \; [W/m^3]$$

or 800 mL per W. However this is excluding the losses in the system, although it also excludes optimisations to the actuator. Most important losses include magnetic losses, which would result in a decrease in both pressure and stroke and the fluid resistance.

4.3. Methodology

The internal pressure of the Ferrofluid depends on the magnetic field and the magnetisation of the ferrofluid. In a simplified formula:

$$\Delta p = \mu_0 \cdot M(H) \cdot (H_2 - H_1) \tag{4.1}$$

Where the magnetisation M is linear dependent on the magnetic field H until the magnetisation saturation, which is a property of the dependent on the type of Ferrofluid. In the case where the magnetic field is high enough, i.e. close to the magnetic source and with high fields then the internal pressure is linear dependent on the magnetic field and thus its losses. This also applies for the stroke, because it is linear dependent to the pressure, therefor the power is quadratic dependent on the losses in the magnetic field. For example a 20 percent loss in magnetic field gives a 36 percent loss in power.

A SolidWorks model is used to have a graphical representation of the geometry used for the calculation, this is seen in figure 4.1. The model will be stackable in the flow direction. Each module consists of two coils on both side of the flow channel if the flow direction is viewed out of the screen. The top and bottom consist of a steel circuit to guide the magnetic field in a direction perpendicular to the flow channel.

The pressure loss due to viscous friction is about linear dependent on both the length of the actuator and the length of the average velocity of the fluid. This relation could be used because the fluid will remain in the laminar regime, i.e. the Reynolds number will be well below 2300. If some common values are used to get a feel for the friction there is seen that the friction could be a few percent of the pressure and thus a few percent of the power, to restricting the flow in such a way that the power drops to only a few percent of the theoretical maximum due to a restriction in flow rate and pressure.

$$\Delta p = \frac{k \cdot \mu \cdot L}{2 \cdot (D_h)^2} \cdot V \tag{4.2}$$

Figure 4.1: SolidWorks render of the pump design with 10 modules

Where k is a factor depending on the shape of the channel, it could vary between 48 and 96, where 48 is a square channel and 96 an infinite wide channel for a channel that is 5 times as wide as it is high this factor is about 75. μ is the dynamic viscosity ranging from of about 0.001 [Pa· s] if water is used as the pumped fluid to 0.1 to 10 [Pa· s] if oil is used. L is the length that the fluid has to travel in the actuator, which should be about 10 [cm], which is about 20 actuators coupled together. D_h is the hydraulic diameter, which is defined as 4 times the area over the perimeter of the channel. This is about $D_h = \frac{2\cdot 0.02\cdot 0.002}{0.022} = 0.36 \cdot 10^{-2}\ [m]$ for the test channel.

$$Re = \frac{D_h \cdot \rho \cdot V}{\mu} \tag{4.3}$$

$$Re_{water} = \frac{D_h \cdot \rho \cdot V}{\mu} = \frac{0.36 \cdot 10^{-2} \cdot 1000 \cdot 2.5}{0.001} = 9000$$

This would mean that in the test channel the water would be turbulent and thus it would not be the right choice due to the high surface roughness of the channel. This could however be tuned with an additive such as sugar to increase dynamic viscosity. This could change the viscosity by a factor 100 this would change the flow regime from turbulent to laminar. If water with sugar is not desirable it could be changed to a low viscosity oil, which could increase viscous friction by a factor 10 or more.

$$Re_{oil} = \frac{D_h \cdot \rho \cdot V}{\mu} = \frac{0.36 \cdot 10^{-2} \cdot 700 \cdot 2.5}{0.1} = 63$$

$$Re_{water+sugar} = \frac{D_h \cdot \rho \cdot V}{\mu} = \frac{0.36 \cdot 10^{-2}) \cdot 1170 \cdot 2.5}{0.0062} = 1.7 \cdot 10^3$$

For a weight percent of about 40% of sugar (sucrose) solution the viscosity is increased [C31] [C32] and it decreased the Reynolds number under the 2300 and thus the flow regime changes to laminar. This is just a quick verification to give an idea of the flow regime. The viscosity could be tuned to meet the specific need by increasing or decreasing the sugar dissolved by reducing the weight percent to 20% the viscosity decreases to 2 [mPa· s], if it is increased to 60% the viscosity increases to 60 [mPa· s]. Therefor even with minor design changes the flow regime could be kept laminar. The pressure drop for the water solution is about:

$$\Delta p_{w+s} = \frac{k \cdot \mu \cdot L}{2 \cdot (D_h)^2} \cdot V = \frac{75 \cdot 0.0062 \cdot 0.1}{2 \cdot (0.36 \cdot 10^{-2})^2} \cdot 2.5 = 4.5 \cdot 10^3\ [Pa]$$

$$\Delta p_{oil} = \frac{k \cdot \mu \cdot L}{2 \cdot (D_h)^2} \cdot V = \frac{75 \cdot 0.1 \cdot 0.1}{2 \cdot (0.36 \cdot 10^{-2})^2} \cdot 2.5 = 2.9 \cdot 10^4 \; [Pa]$$

The pressure drop is about 4.5 percent of the of the total available pressure for the water and sugar. For the oil variant it would be around 29 percent of the total available pressure, thus it reduces the efficiency of the actuator.

The magnetic field is calculated using COMSOL 5.3 in two steps; first the magnetic flux density of the coil is calculated, see figure 4.2. Then this value is average over the core of the coil. The second step is to use this value in the second simulation to calculate the magnetic field inside of the actuator. By splitting this calculation in two steps it is possible to optimize the problem in less computational time. The values used to calculate the magnetic flux in the coil are taken at the absolute limit of what is possible and should therefor be taken as the maximum values that could be achieved by such an actuator. The values used in the introduction of this paper are more reasonable operating limits and thus performance. In figure 4.3 the maximum magnetic field created inside the actuator is seen. With the formula from section background the pressure could be calculated:

$$\Delta p = \mu_0 \cdot M(H) \cdot (H_2 - H_1) = 1.256 \cdot 10^{-6} \cdot (5.2 \cdot 10^4) \cdot (6.4 \cdot 10^5) = 4.2 \cdot 10^4 \; [Pa]$$

The value for magnetic saturation is taken from the EFH3 fluid as it is a well-established and high magnetic saturation ferrofluid.

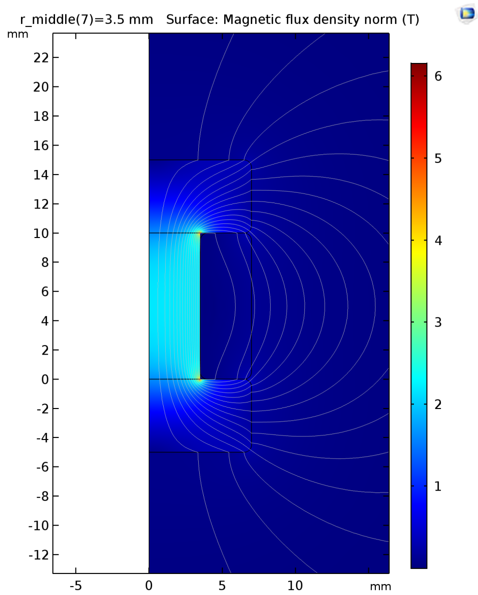

Figure 4.2: COMSOL simulation of the coil around a core

4.4. Results

In the best circumstance a very high pressure could be achieved when the right fluid and the maximum available power is used, which would result in a power density orders of magnitude lower than comparable pneumatic or hydraulic system. That is if the two systems are compared for a single actuator, if a complex system is used, the volume needed for the external part for a hydraulic or pneumatic system are would increase due to the minimum size of these parts. This is where the ferrofluid system is operating at its best, due to the modular design and the lack for the need of any external parts. A maximum pressure of 0.42 bar is achievable but not for a longer sustained period due to heating issues of the coil, at more efficient operating limits at a frequency of around 100 Hz, the pressure is around 0.15 bar where 0.05 bar is lost due to viscous friction in the pumped fluid. Which leaves 0.1 bar with a fluid flow of 10 mL/s per element at 100 Hz for this design.

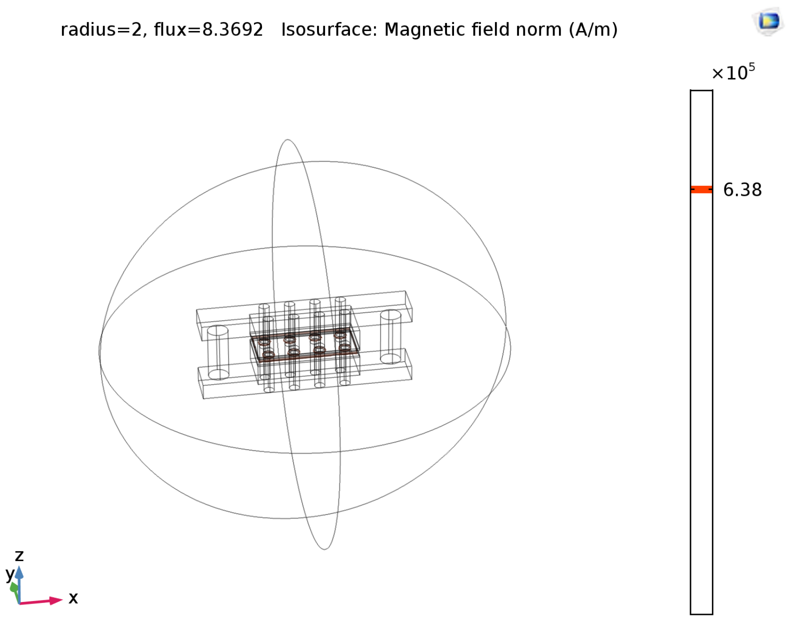

Figure 4.3: COMSOL simulation of the FerroFluid front at maximum usable value of magnetic field of $6.4 \cdot 10^4$ [A/m]

4.5. Discussion

Further optimisation could be performed to increase the field strength, decrease the fluidic loss. With multiple modules connected a more peristaltic motion could be achieved, this might be an interesting option to increase performance. It would be interesting to model both the fluid and the magnetic physics in one simulation to get an idea, due to time constraints this was not possible in this research. This simulation could give more insight in all the losses in the system as described in the section x.

Paper FF References

[C1] D. Brousseau, E. F. Borra, and S. Thibault, “Wavefront correction with a 37-actuator ferrofluid deformable mirror,” *Optics Express*, vol. 15, p. 18190, dec 2007.

[C2] A. S. Boots, L. E. Krijgsman, B. J. de Ruiter, S. G. Lampaert, and J. W. Spronck, “Increasing the load capacity of planar ferrofluid bearings by the addition of ferromagnetic material,” *Tribology International*, vol. 129, pp. 46–54, jan 2019.

[C3] M. De Volder and D. Reynaerts, “Development of a hybrid ferrofluid seal technology for miniature pneumatic and hydraulic actuators,” *Sensors and Actuators A: Physical*, vol. 152, no. 2, pp. 234–240, 2009.

[C4] S. Doganay, A. Turgut, and L. Cetin, “Ferrofluid plug actuation for Micro pumping systems,” *Key Engineering Materials*, vol. 750 KEM, pp. 168–172, 2017.

[C5] H. Hartshorne, C. J. Backhouse, and W. E. Lee, “Ferrofluid-based microchip pump and valve,” *Sensors and Actuators B: Chemical*, vol. 99, pp. 592–600, may 2004.

[C6] E. Uhlmann and N. Bayat, “High Precision Positioning with Ferrofluids as an Active Medium,” *CIRP Annals*, vol. 55, pp. 415–418, jan 2006.

[C7] R. Pérez-Castillejos, J. Plaza, J. Esteve, P. Losantos, M. Acero, C. Cané, and F. Serra-Mestres, “The use of ferrofluids in micromechanics,” *Sensors and Actuators A: Physical*, vol. 84, pp. 176–180, aug 2000.

[C8] N. Tamilarasan, S. Thirumalini, K. Nirmal, K. Ganapathy, K. Murali, and H. Srinath, “Design and simulation of ferrofluid tactile screen for braille interface,” in *International Conference on Robotics and Automation for Humanitarian Applications, RAHA 2016 - Conference Proceedings*, 2017.

[C9] X. Zhang, L. Sun, Y. Yu, and Y. Zhao, “Flexible Ferrofluids: Design and Applications,” *Advanced Materials*, vol. 31, p. 1903497, dec 2019.

[C10] B. Chen, Y. Zhu, J. Zhao, and H. Cai, “Design of a prototype of an adaptive soft robot based on ferrofluid,” in *2015 IEEE International Conference on Robotics and Biomimetics, IEEE-ROBIO 2015*, pp. 511–516, 2015.

[C11] A. Dudu, R. Blue, and D. Uttamchandani, “Miniaturized Ferrofluid Actuated Single Mode Fiber Variable Optical Attenuator,” vol. 7, no. 552, pp. 123–124, 2014.

[C12] M. A. Feizi Chekab and P. Ghadimi, “Ferrofluid appendages: Fluid fins, a numerical investigation on the feasibility of using fluids as shapeable propulsive appendages,” *Journal of Applied Fluid Mechanics*, vol. 10, no. 2, pp. 615–623, 2017.

[C13] Gwan Soo Park and Kang Seo, “A study on the pumping forces of the magnetic fluid linear pump,” *IEEE Transactions on Magnetics*, vol. 39, pp. 1468–1471, may 2003.

[C14] J.-H. Han, J. Tani, J. Qiu, Y. Kohama, and Y. Shindo, “A New Wall Motion Actuator Using Magnetic Fluid and Elastic Membrane for Laminar Flow Control,” *Journal of Intelligent Material Systems and Structures*, vol. 10, pp. 149–154, feb 1999.

[C15] E. S. Keneth, A. R. Epstein, M. S. Harari, R. S. Pierre, S. Magdassi, and S. Bergbreiter, “3D printed ferrofluid based soft actuators,” in *Proceedings - IEEE International Conference on Robotics and Automation*, vol. 2019-May, pp. 7569–7574, Institute of Electrical and Electronics Engineers Inc., may 2019.

[C16] E. Kurtoğlu, A. Bilgin, M. Şeşen, B. Mısırlıoğlu, M. Yıldız, H. F. Y. Acar, and A. Koşar, “Ferrofluid actuation with varying magnetic fields for micropumping applications,” *Microfluidics and Nanofluidics*, vol. 13, pp. 683–694, oct 2012.

[C17] Y. Mitamura and E. Okamoto, “Study of a ferrofluidic actuator for an implantable artificial heart,” *Japanese Journal of Artificial Organs*, vol. 22, no. 3, pp. 748–753, 1993.

[C18] Y. Mitamura and Y. Mori, “Magnetic Fluid-Driven Artificial Hearts,” in *World Congress on Medical Physics and Biomedical Engineering 2006*, pp. 4149–4152, Berlin, Heidelberg: Springer Berlin Heidelberg, 2007.

[C19] Y. Mitamura, T. Wada, and R. Sakai, “A Ferrofluidic Actuator for an Implantable Artificial Heart,” *Artificial Organs*, vol. 16, no. 5, pp. 490–495, 1992.

[C20] T. Schultheis, D. Hoheisel, W. Xiao, L. S. Molella, E. Reithmeier, L. Rissing, and S. Hardt, “Performance of an adaptive liquid microlens controlled by a microcoil actuator,” *Microfluidics and Nanofluidics*, vol. 13, pp. 299–308, sep 2012.

[C21] J. H. Lee, Y. J. Nam, R. Yamane, and M. K. Park, “Position feedback control of a nonmagnetic body levitated in magnetic fluid,” *Journal of Physics: Conference Series*, vol. 149, p. 012107, feb 2009.

[C22] B. Liu, Z. Zhang, J. Yang, J. Yang, and D. Li, “A rotary ferrofluidic vane micropump with C shape baffle,” *Sensors and Actuators, B: Chemical*, vol. 263, pp. 452–458, jun 2018.

[C23] Y. Melikhov, S. J. Lee, D. C. Jiles, D. H. Schmidt, M. D. Porter, and R. Shinar, “Microelectromagnetic ferrofluid-based actuator,” *Journal of Applied Physics*, vol. 93, pp. 8438–8440, may 2003.

[C24] R. Olaru, A. Salceanu, D. Calarasu, and C. Cotae, “Magnetic fluid actuator,” *Sensors and Actuators, A: Physical*, vol. 81, pp. 290–293, apr 2000.

[C25] R. Olaru and C. Petrescu, “SIMPLIFIED APPROACH FOR CALCULATING THE FORCE OF A FERROFLUIDIC ACTUATOR,” *Rev . Roum . Sci . Techn . – Électrotechn . et Énerg*, vol. 53, no. 4, pp. 435–443, 2008.

[C26] R. Olaru, A. Arcire, C. Petrescu, and M.-M. Mihai, “Study of the magnetic force delivered by an actuator with nonlinear ferrofluid and permanent magnets,” *IEEJ Transactions on Electrical and Electronic Engineering*, vol. 12, pp. 24–30, jan 2017.

[C27] R. Olaru, C. Pal, and C. Petrescu, “Current to pressure transducer with magnetic fluid,” *Sensors and Actuators A: Physical*, vol. 91, pp. 150–152, jun 2001.

[C28] C. Petrescu, L. Ferariu, and R. Olaru, “OPTIMIZATION OF FERROFLUID ACTUATOR USING EVOLUTIONARY ALGORITHMS AND FINITE ELEMENT METHOD,” *Rev . Roum . Sci . Techn . – Électrotechn . et Énerg*, vol. 54, no. 1, pp. 77–86, 2009.

[C29] L. Hines, K. Petersen, G. Z. Lum, and M. Sitti, “Soft Actuators for Small-Scale Robotics,” *Advanced Materials*, vol. 29, p. 1603483, apr 2017.

[C30] C. Scherer and A. M. Figueiredo Neto, “Ferrofluids: Properties and Applications,” *Brazilian Journal of Physics*, vol. 35, no. 3A, 2005.

[C31] Engineering ToolBox, “Sugar Solutions - Viscosities.” [online] Available at: https://www.engineeringtoolbox.com/sugar-solutions-dynamic-viscosity-d_1895.html, 2014. [Accessed 23 June 2020].

[C32] Engineering ToolBox, “Density of aqueous solutions of organic substances as sugars and alcohols.” [online] Available at: https://www.engineeringtoolbox.com/density-aqueous-solution-organic-sugar-alcohol-concentration-d_1954.html, 2017. [Accessed 23 June 2020].

5

The feasibility of valve using a magnetorheological fluid as an active medium

Abstract

In this paper magnetorheological fluids will be considered to be used in a channel with a magnetic field applied to it in order have a pressure drop in the channel that is proportional to the magnetic field intensity. This forms the basis for a proportional valve to current with the magnetic field being generated by a coil. The geometry for the valve in this paper is 2 [cm] long, 1 [cm] wide and has the least fluid resistance, a pressure drop of 300 [Pa], while meeting the target pressure drop of 20 [kPa] when the magnetic field is applied, when the height of the channel is 7.6 [mm].

5.1. Introduction

There are many uses of MR fluids in dampers [D1], clutches and brakes [D2]. In these kinds of applications the change in viscosity of MR fluid is used to tune the dampening or transfer the force in the clutches and brakes. When a electromagnet is used in a damper, the dampening is proportional to the current in the coil of the electromagnet. Therefor when the velocity of the fluid is controlled the current is proportional to the current as if it was a valve. To get a rough idea if such a valve would be able to handle about 20 [kPa] of pressure in a small form factor, a MatLab script is used to get an estimate of the performance.

MR Fluids as a smart fluid

This script is based on the reference geometry in figure 5.1. The values for this geometry used to calculated the pressure drops with the script are listed in the table 5.1.

L	=	2	[cm]
w	=	1	[cm]
g	=	0.1-10	[mm]
Q	=	55	[mL/s]

Table 5.1: The flow rate and values used to define the geometry, the gap, g, is variable and used to produce the graphs for this section

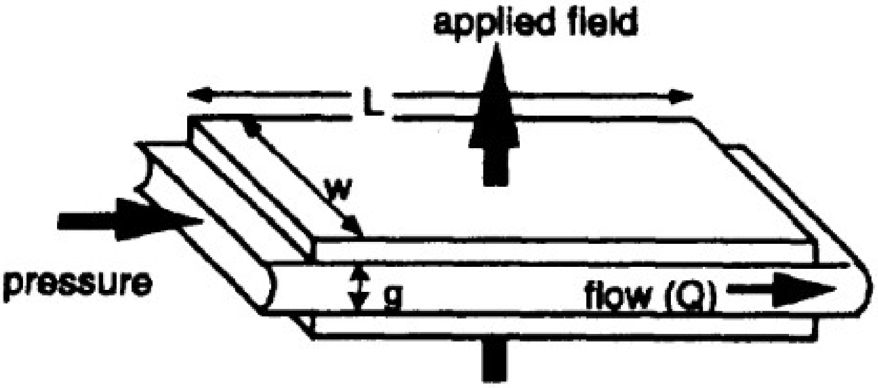

Figure 5.1: Reference geometry for the script, adapted from [D3]

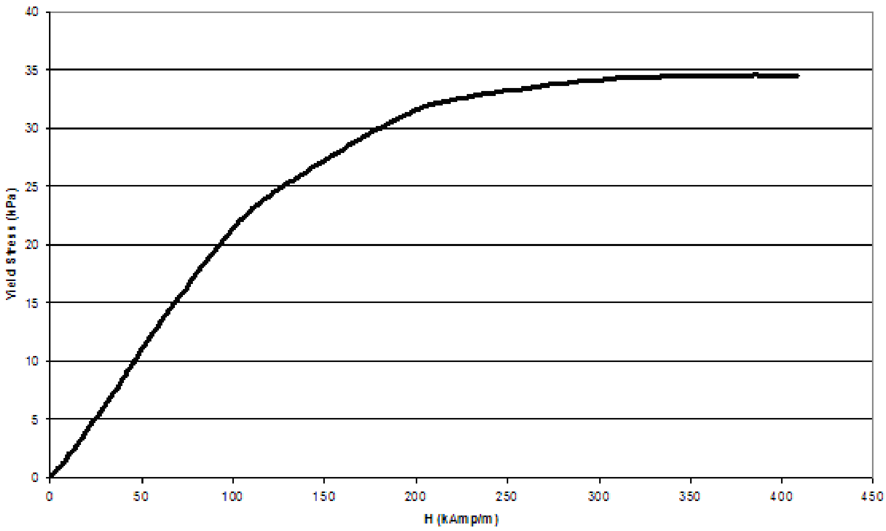

Figure 5.2: Yield strain for different magnetic fields strengths from [D5]

For the script there are five main equations:

$$H = \frac{t_c \cdot w^3 \cdot j}{(w^2 \cdot (g)^2)^{\frac{3}{2}}} \tag{5.1}$$

The first equation is a modified version taken from the course [D4]. Due to the addition of a core the μ_0 will be replaced with $\mu_0 \cdot \mu_r$. Instead of the B field which is used in the equation in these slides it is modified to the H field by dividing both sides by $\mu_0 \cdot \mu_r$. Furthermore the current multiplied with the amount of turns is replace by the current density, j, and the cross section of the coil, $t_c \cdot w$. This is done to have better control over the maximum current density in the coil. t_c is the thickness of the coil and w is the width or diameter of the coil. g is the gap between the coils.

$$\sigma = 36.3 \cdot (1 - e^{-H/109600}) \cdot 1000 \tag{5.2}$$

This equation is a fitted curve based on figure 5.2. The information is based on the MRF-122EG from LORD, a company that produces MR fluids [D5].

$$\Delta P_\sigma = \frac{C \cdot \sigma \cdot L}{g} \tag{5.3}$$

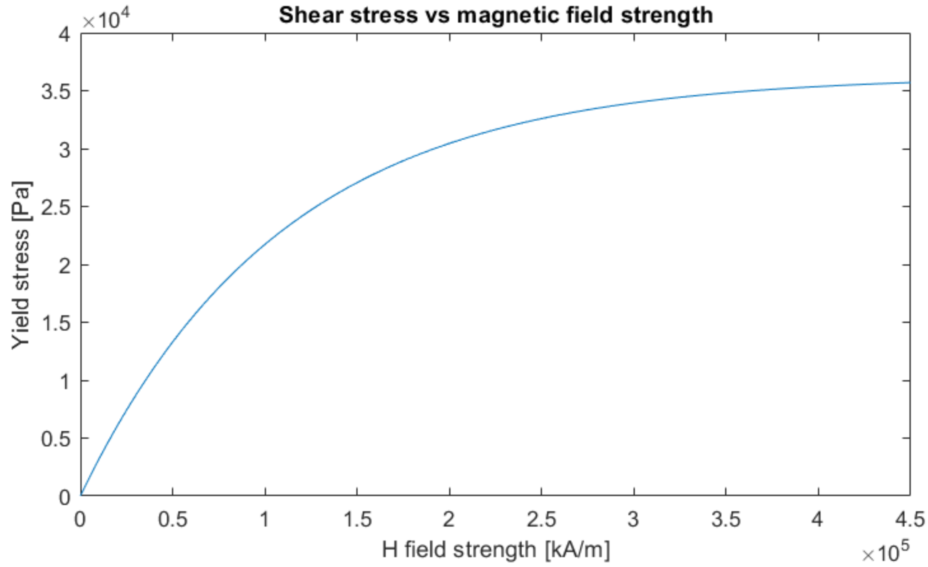

Figure 5.3: MatLab generated figure of the curve fitted equation of the figure 5.2 for the yield stress in the MRF. Note that the curves are similar except for the high fields, which are not used in the script.

$$\Delta P_\eta = \frac{12 \cdot \eta \cdot Q \cdot L}{(g)^3 \cdot w} \tag{5.4}$$

Both the equation for the pressure drop related to the increase in viscosity of the MR fluids by the magnetic field ΔP_σ and due to standard viscosity of the fluid, ΔP_η, are taken from [D3], the geometry of the channel is dictated by the width, w, the height of the channel or the gap, g, and the length of the channel, L. C is a factor that ranges from 2 when the two values of the pressure drops are close and 3 if the pressure drop of the ΔP_σ is way bigger, about 100 times, than the pressure drop due to viscosity, ΔP_η. For this script it is chosen to let C be 2, to underestimate the pressure drop and cover some of the potential losses. The flow rate is needed for the viscous losses is represented with Q, this value is an estimation for quarter of the maximum flow that is needed to aid the heart as presented in the paper in chapter 2. This is an over estimation of the flow rate to show that the valve has a higher potential.

$$P_E = 2 \cdot (j \cdot C_{cf})^2 \cdot \rho_{Cu} \cdot t_c \cdot \pi \cdot (0.5 \cdot w)^2 \tag{5.5}$$

To calculate the heat generated per second, P_E, with a current density j in the 2 coils is multiplied with the current factor, C_{cf}, i.e. the amount of current that is needed to get to the pressure difference of the magnetic field P_σ to the required pressure of 20 [kPa]. The total current then is squared and multiplied with the resistivity of copper ρ_{Cu} and the total volume of copper, $t_c \cdot \pi \cdot (0.5 \cdot w)^2$

Results

The pressure drop of the MRF is used to as a valve, therefor the maximum pressure drop needed is the pressure estimate from the (had paper) at 20 [kPa]. The estimation for the power needed is based on this metric. In figure 5.4 there is seen that the gap in this geometry should be smaller than 7.6 [mm] in order to have the required 20 [kPa]. At 1.8 [mm] and smaller the pressure drop of the viscous friction is as high as the pressure required. This would cause no problems if the pump could supply enough pressure, but it is a trade-off between power used by the pump and the power used by the MRF valve. The power needed, decreases by two decades per decade with a smaller gap due to the maximum yield stress decreasing linearly with the gap size and the pressure drop both decreasing linearly with the gap and increasing with the maximum yield stress, as seen in figure 5.5. When the gap size approaches 10 [mm], the decrease in power needed to achieve the required pressure drop is even higher per decade, due to the non linear relation of the maximum yield stress with the magnetic field.

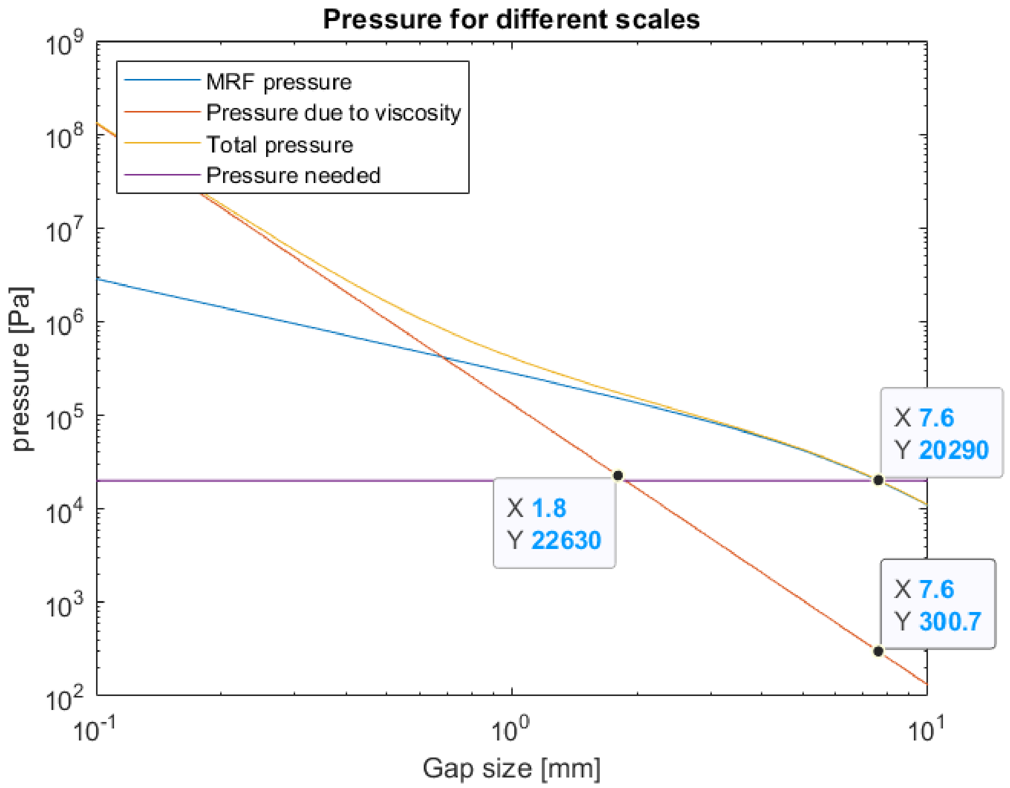

Figure 5.4: Estimated pressure at centimetre scale, figure generated by MatLab script

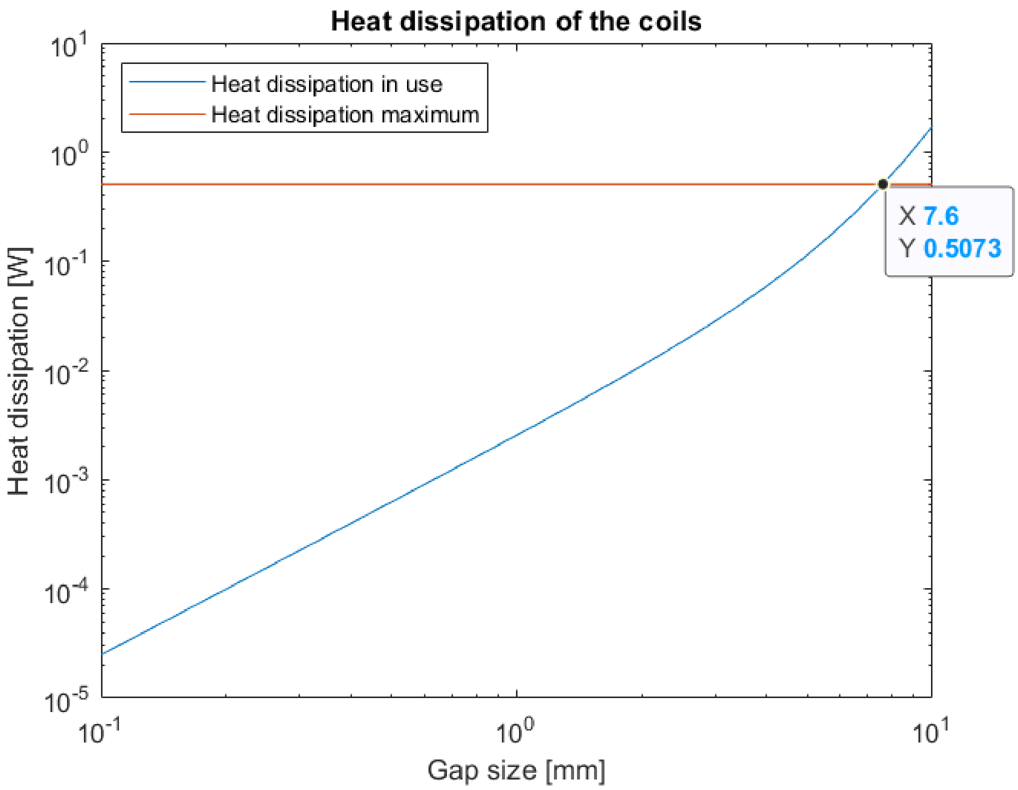

Figure 5.5: Scaled power to match target pressure of 20 kPa, figure generated by MatLab script

5.2. Conclusion

For the geometry defined in table 5.1, the gap of 7.6 [mm] will result in the least viscous friction while the required pressure drop of 20 [kPa] is achieved. When the gap is made smaller the power consumption goes down by about two decades per decade if the pressure drop of 20 [kPa] is maintained. Therefor it can be concluded that the MR fluid based continuous variable valve, would be viable as direct current to pressure drop transducer with a 20 [kPa] pressure drop with the minimum viscous losses. The pressure drop could be increased, but the viscous friction will rise faster with about 3 decades per decade for the gap size versus the pressure drop due to the applied magnetic field with 2 decades per decade for the gap size.

5.3. Discussion and recommendations

The field that is calculated is based on two thin coils with no core. A possibly more accurate way of modelling this could be by using a electric circuit analogy. By using this method, a better understanding of the different losses and fields in the different parts of the circuit would be possible.
At this moment the parameters of the MRF are taken from the spec sheets of LORD. A better way of finding these numbers is to measure these values in a more comparable use case to the valve. The second advantage is that a more elaborate model like the Herschel-Buckley model could be implemented and thus a more accurate pressure drop could be estimated.

Paper MRF References

[D1] N. P. Sherje and Deshmukh S. V., "Characterization of Magnetorheological Fluid For Damper In Automobile Suspension," *International Journal of Mechanical Engineering and Technology*, vol. 7, no. 4, pp. 75–84.

[D2] J. de Vicente, D. J. Klingenberg, and R. Hidalgo-Alvarez, "Magnetorheological fluids: a review," *Soft Matter*, vol. 7, p. 3701, apr 2011.

[D3] M. R. Jolly, J. W. Bender, and J. D. Carlson, "Properties and Applications of Commercial Magnetorheological Fluids," *Journal of Intelligent Material Systems and Structures*, vol. 10, pp. 5–13, jan 1999.

[D4] R. Nave, "Field on Axis of Current Loop." [online] Available at: http://hyperphysics.phy-astr.gsu.edu/hbase/magnetic/curloo.html#c3.

[D5] LORD, "MRF-122EG Magneto-Rheological Fluid." [online] Available at: http://www.lordmrstore.com/lord-mr-products/mrf-122eg-magneto-rheological-fluid.

6
Conclusion

This work investigated the feasibility of a novel actuation unit for the purpose of creating a heart assist device. First, it studied the requirements that are need for a heart assist device, secondly performance of the ferrofluid pump and lastly theoretical side of the MR fluid based valve.

First, the requirements of a actuation unit for a heart assist device were established using a self-written MatLab script; The average heart, at 70 percent of its original muscle force needed for rest heart rate, needs about 20 [kPa] for a 5 [mm] thick shell to maintain a normal rest heart rate. This shell should be able to compress 20 percent of its maximum size. This outcome is an estimation to aid in the selection process of the actuators. Furthermore, 3 [cm] of distance perpendicular to the hearts surface is available for the design of the actuation units. This should include all the components of such a system because it would be beneficial to avoid any need for a opening in the chest, for cooling or otherwise. Finally the cooling capacity is estimated by assuming that 5 percent of the blood flow, which is the flow around the heart, carries the heat flux of the system. This amounts to 10 [W] of cooling power.

Next, a ferrofluid pump was investigated to evaluate its feasibility for pump for a segment of a heart assist device. A design was purposed to act as a test bed and used to theoretically identify the performance in pressure and flow rate. This solution would be able to achieve a pressure of 10 [kPa] with a flow rate of 10 [mL/s] while keeping the pump bound in a 4 [cm] by 2 [cm] by 1 [cm] volume. This could be useful in systems where ferrofluid needs to be pumped around, e.g. a sliding platform moving over ferrofluid bearings that needs to maintain a certain pressure. Unfortunately it will not be sufficient to be used to actuate a single segment and therefor it is not considered as a solution for the heart assist device.

Finally, magneto rheological fluids were investigated to be used as an active fluid to design in inline variable valve to distribute the fluids in the different segments of the heart assist device. This solution could be used to step a pressure down to the required pressure by having a pressure drop over the valve that is controllable with a magnetic field. A pressure drop of 20 [kPa] is achievable with commercially available MR fluids in a 1 [cm] long in flow direction by 7.6 [mm] high channel. The width of the channel will not directly affect the pressure drop related to the change in viscosity of the MR fluid, but it will have an effect on the pressure drop due to viscous friction. With this 7.6 [mm] height of the channel and a width of 2 [cm] about 0.5 [W] is used to achieve the 20 [kPa] pressure drop while the viscous friction related pressure drop is less than 0.1 [kPa] at full actuation speed. Therefor this solution would be suitable as an actuation method and it would be recommended to be further developed in a heart assist device. A good place to start would be the experimental validation of the theory gathered in this thesis.

7 Recommendations

The recommendations consists of three subjects; the requirements for the project, the ferrofluid based pump and lastly the variable valve using the MR fluid's changeability in viscosity. The different subject will have some general remarks about what could have been done differently in this research followed by a list of recommendations for further research.
The requirements for the heart assist device were obtained using a MatLab script. This script combined parameters of medical research and textbooks with a basic model to convert these medical values to an estimation for the mechanical parameters. This model was only used for this estimation and should be adjusted and compared to measurements of the heart before a actuator could be designed using the values from this script. The model could be improved on several aspects:

- A numerical integration scheme could be applied to get better and faster results instead of the fixed time step in the current script.
- The shape of the computational version of the heart used in the script is a cylinder, this could be changed into a more realistic shape to obtain a better match with a real heart.
  - An other addition to the model could be to include the direction of the fibres to get a more realistic idea of the motion of the heart.
- An inverse version of the model used in the script could be used to find the mechanical parameters of the heart, like force of the heart muscle over time.
  - A version of this inverse model, where the heart is segmented into discrete sections, could be developed into the basis to test actuation done by an actuator to the heart.
  - Integration of this script with a 3d ultrasound could be developed into a way of extracting data from a patient to form a model of the patient's heart using a method that is already used in hospitals.
- Lastly, there is the need for further research to find how much heat could be dissipated in the region around the heart.

Next, the ferrofluid pump was not suitable to be used in the intended purpose for this research, that however does not mean the concept of ferrofluid being pumped using an magnetic field could not be used in other applications. There are several recommendations to make the model more accurate;

- A full finite element simulation could be done including all the ferrofluid and all of the components needed, where the ferrofluid is still used as a fluid. Due to the inclusion of the fluidic behaviour this would only be recommended for a late stage in a design process.
- For a quick estimation a lumped magnetic model could be used, with the estimation of the pressure of the ferrofluid using the formula described in the paper about ferrofluids in chapter 4. An application where the pump based on ferrofluids would be an interesting concept is a sliding platform on ferrofluid bearings. The ferrofluid pump could control the height of the platform by supplying ferrofluid or pumping it away.

Finally, the valve based on MR fluid was viable to be used as an active element in a heart assist application. There are some recommendations for further research:

- For a better estimation of the magnetic field on could use a lumped magnetic model, to have a more accurate number for the magnetic field strength. This could be used to quickly calculate the difference between setups, e.g. using a different amount of turns in the coils, a change in layout or different materials for the magnetic circuit.
- The valve seems very promising in theory and should be tested experimentally to see if the performance of the valve agrees with the theory.
  - One of the theories that should be tested is if the Bingham model holds true, or that another model is necessary, e.g. the Herschel-Buckley model.
- Another test that would be useful is to test the increase of the maximum pressure drop with the decrease of the gap size, or height of the channel. According to this thesis; for every decade of decrease of channel height the pressure drop will go up by two decades, but the viscous friction will go up by three decades. This will make the device more sensitive to manufacturing errors, i.e. a mismatch between two different valves.
- Lastly, another application where the MR fluid based valve could be useful is in simulators where it could add damping to the controls for the human to machine interaction to match the real world in a simulation.